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AUTHOR Richardson, Charles; Simmons, Roger W.
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Electromyographic and Neuromuscular Force Patterns Associated with Unexpectedly Loaded Rapid Limb Movements

Charles Richardson and Roger W. Simmons

San Diego State University

Author:

Charles Richardson

Roger W. Simmons

Human Performance Laboratory, PG. 214,

San Diego State University

San Diego, CA 92182

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SUMMARY Bi-articular, unidirectional arm movements were studied to evaluate the electromyographic (EMG) and neuromuscular force patterns that occur when a limb is unexpectedly perturbed. A series of training trials were continued with a control load spring attached to the apparatus until a pre-specified criterion for learning was attained. The limb was then unexpectedly loaded with one of four test load springs. Examination of the integrated EMG records revealed a coactivation pattern of neuromuscular activity during a major part of the movement. Analysis of applied force data supported the notion of sustained agonist activity. Reciprocal innervation and coactivation patterns were considered in terms of a common motor control system as proposed by the equilibrium point control hypothesis. Manifestation of coactivation rather than reciprocation in the present study was discussed in terms of the establishment of 'mental set'.

Studies utilising electromyographic (EMG) techniques to investigate the role of motor programmes in controlling moderate to high-speed limb movements consistently report a reciprocal pattern of neuromuscular activity[1-7]. This so called triphasic pattern, which consists of two discrete bursts of agonist activity separated by a period of antagonist activity, has been observed in healthy humans[8], athetoid individuals[9] and intact as well as deafferented animal preparations[10].

Despite a high frequency of citation in the literature a triphasic pattern is not the neuromuscular pattern predicted by some models of motor control currently used to explain the production of rapid limb movements. For example, the mass-spring model[11, 12] has the length-tension characteristics of a muscle assuming properties normally associated with mechanical springs. The length, and therefore the tension developed in the agonist and antagonist, is determined by pre-setting the level of alpha-motorneuron activity to both muscle groups at a central level. Final limb position occurs at the point where the torque developed between the two muscle groups algebraically sums to zero. The unexpected application of an inertial or torque load does not affect the degree of alpha-motorneuron activity. Therefore, final limb position or the equilibrium point is achieved independently of load and without the need for

afferent input[13, 14]. However, a critical element of the model is the simultaneous innervation of the agonist and antagonist groups from the time of movement initiation to the moment when the limb is stabilised at the equilibrium point. In terms of electromyographic activity, evidence for the mass-spring model would appear as a pattern of neuromuscular coactivation and not a triphasic pattern.

The inconsistency between theoretical modelling and the observed EMG activity has been resolved by associating triphasic activity with rapid limb movements and coactivtion with tasks involving the stabilisation of body segments, as required for the maintenance of posture[15, 16]. However, at least two sets of data challenge this restricted motor application. First, Fel'dman[17] has demonstrated simultaneous agonist and antagonist activity during the tonic phase of a rapid limb displacement task. However, the limb was unloaded and so the type of EMG pattern associated with loaded movements remains unclear.

Second, the force applied to the apparatus during a linear movement varies directly with the size of a spring load, which suggests that the agonist is active in overcoming the spring resistance throughout the movement[18]. In addition, the antagonist should be activated to facilitate stabilisation of the movement at the target area. Thus, at some point during the movement coactivation must have resulted. This

conclusion is tenuous however, as it is based on indirect observations of applied force data.

The purpose of the present experiment was to investigate the EMG activity and neuromuscular force patterns resulting from unexpected perturbation of a bi-articular rapid limb movement. Consistent with the mass-spring hypothesis, it was predicted that a neuromuscular pattern of coactivation, rather than triphasic activity, would be observed during a substantial portion of the movement. Limb displacement and applied force data were also considered, to ensure that the behavioural response outcomes of the present study did not significantly differ from data reported in previous experiments. This procedure not only expanded the scope of the experiment, but also allowed the investigator to eliminate the possibilty that the observed EMG patterns, such as coactivation, were not a manifestation of atypical responses.

Methods

Twelve right-hand dominant males moved a near frictionless slide along a linear trackway. A 30 cm linear potentiometer and a Statham AD-3 linear accelerometer were attached to the rear of the slide. The electrical analog signals produced by both units (limb displacement and acceleration respectively), were recorded on a Honeywell

906-B Visicorder using two 3000 Hz critically-damped mirror galvanometers at a paper speed of 63.5 cm/sec.

A 35 cm load spring was anchored to the apparatus base board and the rear of the slide via two small metal hooks. The load spring rested in a teflon cradle located between the two hooks. Five load springs were constructed of wire of varying diameters to obtain different load characteristics with constant spring length. The five springs were calibrated at different lengths of extension but were labelled as 1.0, 1.6, 2.0, 2.7 and 4.5 kg, on the basis of the tension developed at 20 cm of extension. Subjects were randomly assigned to one of four groups with each group using the same control load spring (1.0 kg) but a different test load spring. Groups 1, 2, 3 and 4 were assigned test loads of 1.6, 2.0, 2.7, and 4.5 kg respectively.

Muscle action potentials were recorded using Beckman bipolar surface electrodes connected to a Beckman dynagraph. Two pairs of electrodes were placed vertically over the superior fibres of the right pectoralis major (PM) muscle (agonist) and two electrode pairs over the posterior fibres of the right deltoid (DM) muscle (antagonist). A ground electrode was fixed over the left external oblique muscle. The interelectrode impedance was below 5,000 ohms on all measured trials.

After several practice trials, all subjects were blindfolded and followed the same protocol. The subject self-initiated a rapid (less than 200 ms) right-to-left 20 cm movement towards a 2 cm wide target area. The movement was loaded with a 1.0 kg control spring. Knowledge of results concerning the temporal and spatial errors was provided after each training trial. Trials were repeated until five consecutive movements were observed with the movement terminating in the target area in less than 200 ms.

The fifth correct training trial was followed by a single measured trial during which the experimenter removed the 1.0 kg load spring from the apparatus and either immediately returned the spring to the slide, or replaced it with the appropriate test load spring. The subject was instructed to attempt the same movement as executed during the five successful training trials. Procedures for interchanging spring loads were identical on all measured trials and prevented the subject from knowing which spring was attached to the apparatus. Upon completion of the measured trial, the control load was re-applied to the slide, if not already in place, in preparation for the next set of training trials.

This sequence of training trials followed by a single measured trial was completed eight times by each subject during a single testing session. Four randomly determined

measured trials were completed with a control load spring and four trials with the test load spring attached to the apparatus.

Results

Trials to criterion. The number of training trials completed prior to each measured trial was analysed using Wilcoxon's test for dependent samples. No significant differences were detected which indicated the amount of practice experienced by each subject did not differentially affect response outcomes.

Movement time. The temporal patterns associated with EMG activity are related to total movement time.[8] To ensure that movement time did not vary across load conditions, the mean movement time for the four control load and four test load trials was measured to the nearest millisecond and statistically analysed. No significant main effects or interactions were indicated by the analysis.

Movement distance. Limb displacement was measured to the nearest millimeter. Statistical analysis of the mean algebraic error for the four control and test load trials presented in Table 1, resulted in a significant group by load interaction [$F(3,8) = 5.25$, $p < 0.05$]. Additional analysis of the interaction indicated that during control

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Insert Table 1 about here

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load trials the algebraic error for all groups fell within the target area (20 ± 1 cm). Unexpected loading however, resulted in significant underestimation of the target area for groups 1, 2, 3 and 4 [$F(1,16) = 19.2$, 54.7, 71.4 and 94.0, $p < 0.0001$ respectively]. The degree of undershoot varied directly with the size of the test load [$F(3,16) = 16.33$, $p < 0.001$].

Applied force. The force applied to the hand/apparatus system is given by the equation

$$Fa = kcx + b\dot{x} + m\ddot{x}$$

where Fa is operationally defined as the force applied by the subject to the apparatus, x is the linear displacement of the trolley, and kc is the stiffness coefficient associated with each spring load (0.05, 0.08, 0.1, 0.135 and 0.255 kg/cm for the 1.0, 1.6, 2.0, 2.7 and 4.5 kg loads respectively). The viscous damping coefficient b was negligible and was not included in the calculations. Inertia is represented by $m \approx 0.03$ kg sec^2/m. Acceleration values ($\ddot{x}$) were determined by sampling the analog acceleration record every 10 ms. Velocity values are represented by $\dot{x}$.

Analysis of the applied force data, presented in Figure 1, revealed a significant group by load interaction [$F(3,8) = 8.74$, $p < .001$]. The applied force values observed during

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Insert Figure 1 about here

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the control load trials were statistically equivalent for each sampled time interval for groups 1 through 4, but applied force increased significantly for all groups when the movement was perturbed by the test load. A within group analysis indicated the initial segment of the control and test load applied force curves to be similar, but significant separation did occur between the two sets of curves after 80 ms for groups 1 and 2, 60 ms for group 3 and 50 ms for group 4. It was concluded that differences in applied force within a group were a function of the size of the control and test spring loads.

Electromyographic activity. The integrated EMG signal for both agonist and antagonist muscles during control and test load conditions for subject 7 assigned to group 1 (1.6 kg) is presented in Figure 2. Although unique characteristics

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Insert Figure 2 about here

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are a part of this record, it contains three distinct features that were observed in the EMG records for individuals assigned to all groups.

First, there was a rapid rise in agonist (PM) activity with maximal electrical activity occurring approximately at the time of movement initiation. This agonist activity was sustained throughout the course of the movement but at reduced levels of amplitude. Second, the antagonist (PD) was activated shortly after movement initiation and remained active, again at reduced levels of intensity, beyond the point of movement termination. The result was a pattern of coactivation during a substantial portion of the movement. Finally, the waveforms between control and test load conditions and across groups were very similar. This effect is best observed when indivdual records are averaged within a group, as represented in Figure 3. Using the averged EMG records as a data base, the role of neuromuscular phasing during coactivation was clarified by examining the time relations between the onset of the agonist and antagonist

Insert Figure 3 about here

activity. Statistical analysis indicated an invariant absolute time difference across all conditions [$F(3,8) = 1.69$, $p > .01$]. It was concluded that phase time did not vary with load.

Discussion

The present experiment was primarily concerned with testing a prediction of the mass-spring theory of motor control; that is, a bi-articular spring-loaded rapid limb movement is subserved by a coactivation pattern of neuromuscular activity. Data from individual and group (averged) EMG records supported this prediction for the four levels of loading utilised in the experiment.

However, other predictions of the mass-spring model are not consistent with the limb displacement or applied force data. The model stipulates a central control system responsible for pre-setting neuromuscular innervation levels prior to movement.[12] Accordingly, the final limb position during test load trials should be located at a point where the developed force is equal to that generated during control load trials (1.0 kg). Specifically, final limb displacement values of 12.5, 10.0, 7.4 and 4.5 cm would be

expected for test loads of 1.6, 2.0, 2.7 and 4.5 kg, respectively. The observed values for displacement (Table 1) and applied force (Figure 1) exceeded the predicted locations and force levels in all groups. This result implies an ongoing modification of the centrally determined movement parameters that is not predicted by the mass-spring model. Therefore, another theoretical description is required to explain the limb displacement and applied force data, together with the EMG results.

Unlike the mass-spring theory which stipulates invariant alpha-motorneuron activity, the equilibrium point control hypothesis recognises that alpha-motorneuron activity is load dependent and reflects a mixture of efferent and afferent information.[19] (See Fel'dman[20, 21] for a complete discussion of the differences between the two models). Consequently, this model predicts modification of the response during the movement while retaining the notion of coactivation. In both respects, the model is consistent with the observed results.

The equilibrium point control hypothesis also provides an appropriate theoretical description for explaining the difference between the present EMG pattern of coactivation and those studies reporting triphasic activity. In contrast to other models of motor control, the equilibrium point control concept proposes that rhythmic and unidirectional

movements consist of the superposition of both reciprocal (triphasic) and coactivating programmes. The emergence of one neuromuscular pattern over another may be attributable to the establishment of different 'mental sets' or strategies[22] developed in response to prior instruction[23, 24], the degree of previous experience[25] or the types of load used during experimentation.

A response to an unloaded or inertia loaded movement may be best accomplished through a strategy involving precise limb deceleration by a controlled neural pulse to the antagonist. Positional accuracy of the movement would depend on the antagonist duplicating the torque developed by the agonist at a specific time following movement initiation and scaled according to the anticipated load.[26] The resulting neuromuscular activity would represent a reciprocal innervation pattern.

During a variable loaded movement, spatial accuracy may result from a strategy which regulates joint stiffness by coactivating the musculature in repsonse to the varying resistance. In this example, the degree of joint stiffness is the result of an interaction between the intrinsic muscle non-linearities and autogenetic reflexes[27] which occurs independent of joint angle.

The adoption of specific strategies when encountering various types of movement perturbation not only accounts for

the differences in neuromuscular activity observed in the present and previous studies, but may have implications for the different control systems noted for ramp and ballistic movements. However, the temptation to assign several classes of movement to a single controlling mechanism should be resisted. Based on recent information[28], it is reasonable to suggest that several variables are used in control of a movement. Thus it may prove more useful to investigate the interaction of these variables, such as the superimposition of reciprocal and coactivating programmes suggested by Fel'dman[19], if a clearer understanding of how we control movements is to emerge.

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Table 1

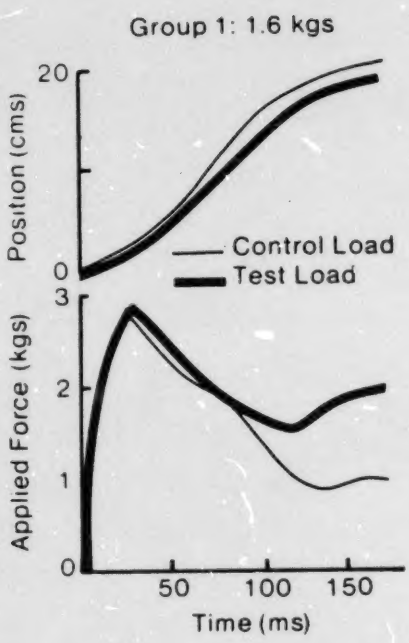
Mean Algebraic Error for a Rapid Limb Movement Unexpectedly Pertured by a Spring Load*

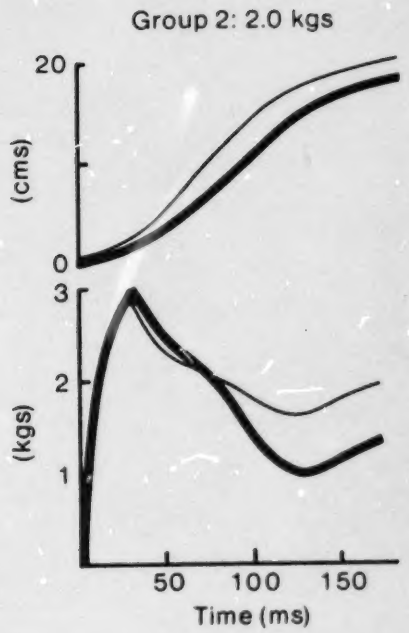
Load Condition	Group 1(1.6kg)		2(2.0kg)		3(2.7kg)		4(4.5kg)		
	M	SD	M	SD	M	SD	M	SD	Row M
Control Load	-0.48	1.4	-0.02	1.7	-0.52	1.2	0.82	1.2	0.05
Test Load	-1.59	2.7	-2.38	1.0	-2.87	1.3	-5.28	2.9	-3.03
Column M	-1.04		-1.20		-1.70		-2.23		

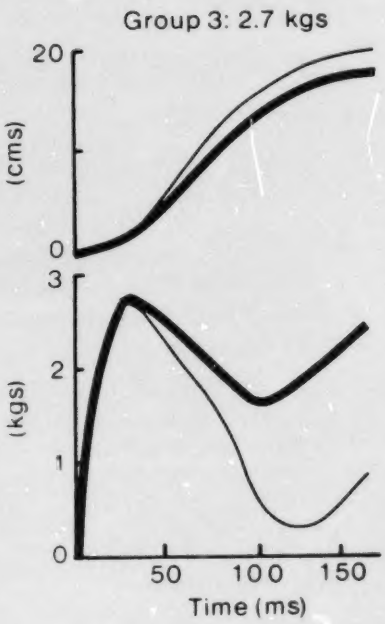
*Data are expressed in centimeters

Figure Caption

Figure 1. Mean applied force and movement displacement values during control (1.0kg) and test load (1.6, 2.0, 2.7 and 4.5kg) trials as a function of movement time.


Group 1: 1.6 kgs
Position (cms)
20
0
Control Load
Test Load
Applied Force (kgs)
3
2
1
0
50
100
150
Time (ms)


Group 2: 2.0 kgs
(cms)
20
0
(kgs)
3
2
1
50
100
150
Time (ms)


Group 3: 2.7 kgs
(cms)
20
0
(kgs)
3
2
1
0
50
100
150
Time (ms)

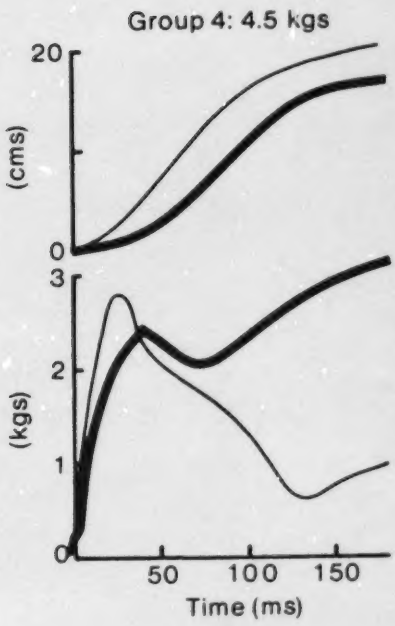
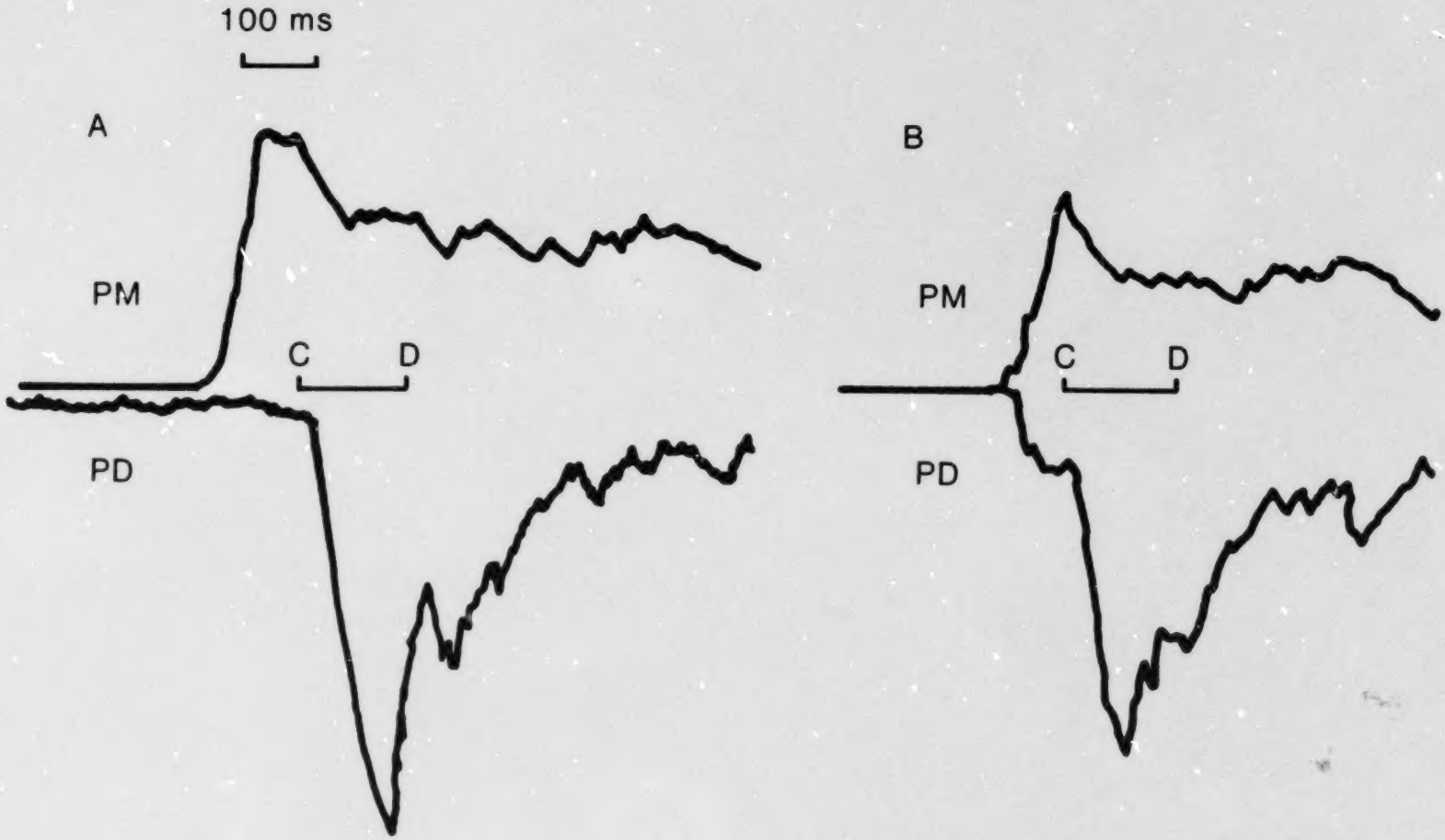

Group 4: 4.5 kgs
(cms)
20
0
(kgs)
3
2
1
0
50
100
150
Time (ms)

Figure Caption

Figure 2. Representative EMG and displacement records for a subject assigned to group 1. A is the EMG record for a single control load (1.0kg) trial and B the EMG record for the test load (1.6kg) trial. C and D represent the points of movement initiation and termination respectively. PM-Pectoralis major; PD-Posterior deltoid. The EMG record for PD has been inverted to facilitate a time course comparison with PM.

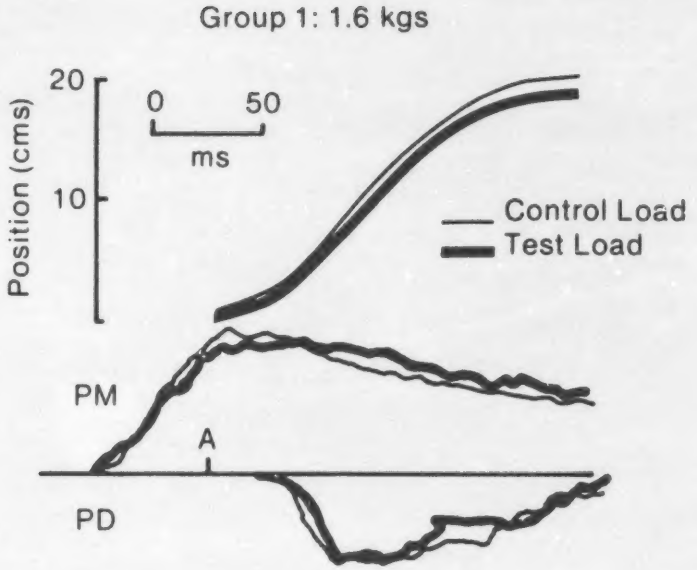


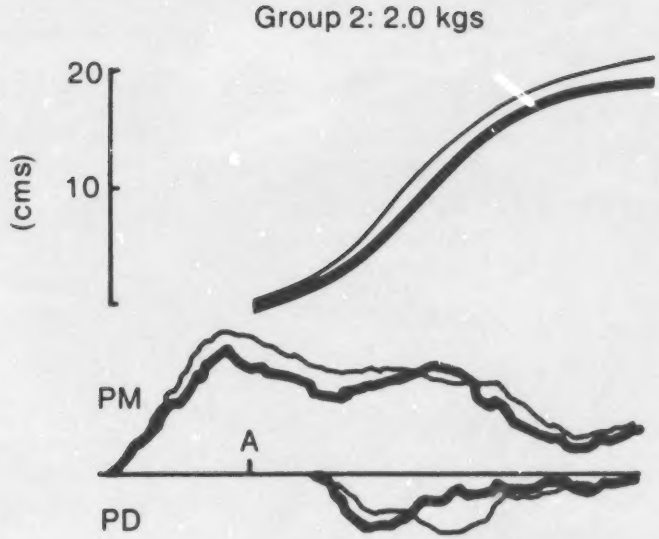
25

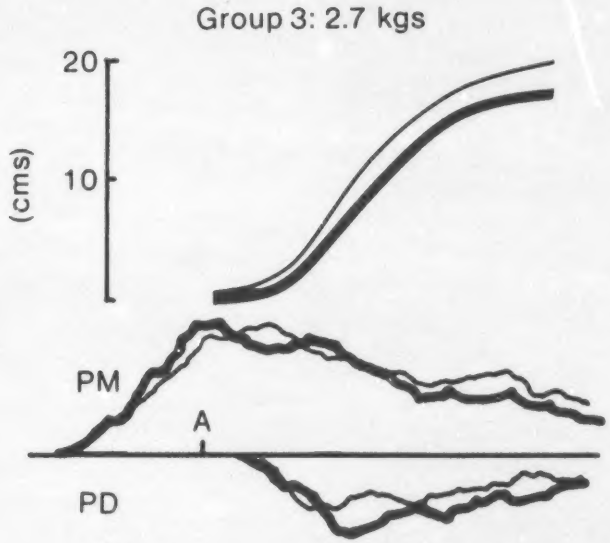
26

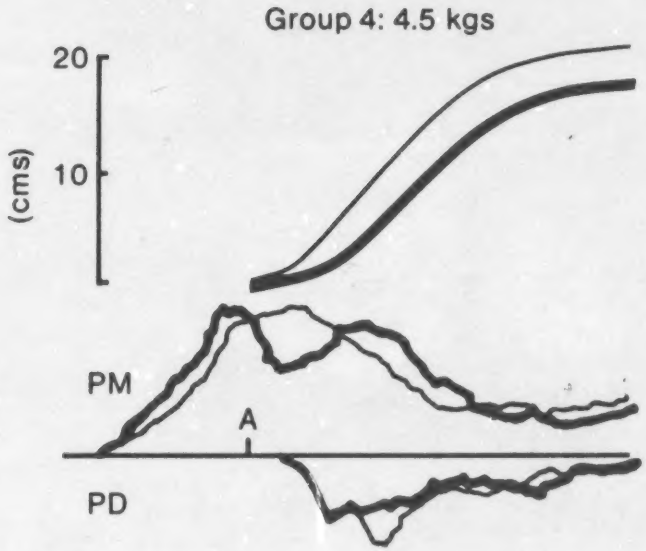
Figure Caption

Figure 3. Averaged EMG activity associated with control (1.0kg) and test load (1.6, 2.0, 2.7 and 4.5kg) trials as a function of movement time and limb displacement. PM-Pectoralis major; PD- Posterior deltoid. The EMG records for the PD have been inverted to facilitate comparison with the time course of PM.


Group 1: 1.6 kgs
20
10
Position (cms)
0
50
ms
Control Load
Test Load
PM
A
PD


Group 2: 2.0 kgs
20
10
(cms)
PM
A
PD


Group 3: 2.7 kgs
20
10
(cms)
PM
A
PD


Group 4: 4.5 kgs
20
10
(cms)
PM
A
PD